Quantifying the Effects of Blood Pressure Changes on Ballistocardiogram Signals

Abdul Q. Javaid, Hakan Töreyin and Omer T. Inan

Abstract— **Quantifying the effects of blood pressure (BP) changes on the ballistocardiogram (BCG) signal shape and features can potentially improve the understanding of this mechanical modality of cardiovascular sensing. BCG, a measure of body movements caused by ejection of blood into the vasculature, has recently re-emerged as a promising method for trending cardiac output and myocardial contractility. Although recent research has shown that the BCG waveform has the potential to be used as a viable proximal timing reference for measuring pulse transit time (PTT) and indirectly BP, it has not been deeply explored for direct estimation of BP. In this paper, we posit that the BCG signal contains features corresponding to changes in BP. To further investigate this hypothesis, BCG waveforms were measured using a modified-weighing scale from 14 subjects performing an isometric hand-grip challenge in a seated position. The energy in the latter half of the BCG heartbeat was estimated using polynomial fitting and interpolation methods. The results indicate that an increase in mean arterial pressure (MAP) or a decrease in PTT manifests itself in the form of high amplitude oscillations following the main peak (J-peak) in a BCG heartbeat, thus elucidating the mechanisms behind these oscillations and also potentially improving the breadth of data that can be sensed using BCG signals.**

I. Introduction

Hypertension (high blood pressure) is an important cardiovascular risk factor. Nearly half of all hypertensives have uncontrollable blood pressure [1], thus requiring easy-to-use and robust methods for continuous monitoring. However, current techniques to monitor blood pressure (BP)—catheterization [2], volume clamping, auscultation [3], oscillometry [4], and tonometry [5]—are invasive or obtrusive and cannot provide continuous monitoring. Moreover, some of these methods require trained personnel to conduct the test. Hence, there is a pressing need for systems which can provide continuous and ambulatory monitoring of BP, thus allowing for real-time changes in medication for better management of hypertension and improving the quality of life.

Research has shown that pulse transit time (PTT)—the amount of time it takes a pressure wave to travel between two points on arterial tree—can be measured from proximal and distal waveforms and provide the basis for convenient cuff-less BP monitoring [6]. PTT is inversely proportional to BP, i.e., $\text{PTT}^{-1} \propto \text{BP}$. The PTT-BP relationship has been extensively studied in existing literature [7]. However, PTT measurements are vulnerable to errors due to changes in elasticity of peripheral arteries with aging and disease. Hence, central arteries provide a better location for accurate estimation of PTT [7].

Recently, the ballistocardiogram (BCG) signal, a measure of the reactionary forces of the body to the flow of blood in the vasculature, has re-gained attention as a useful technique for continuous and robust estimation of mechanical parameters of cardiovascular function. It has been shown that the BCG sensors have the potential to estimate flow-related hemodynamic parameters [8]. Recent research has shown that the time interval between the ECG R-peak and the highest peak in the BCG signal (J-peak), called the RJ-interval, is directly proportional to pre-ejection period (PEP) —a surrogate measure of cardiac contractility [9]. Some of the earlier research has also shown that a feature in the BCG signal, the K-peak, can be used to assess changes in BP [10]. Since the BCG methodology can provide an estimate of PEP, it can be combined with other sensors (e.g., photoplethysmography, PPG), which measure pulse-arrival time (PAT), to accurately estimate PTT and hence BP [6], [11].

However, researchers have not deeply investigated the changes in BCG morphology in association with the changes in BP. Specifically, along with providing a viable proximal timing reference for PTT calculation, features in the BCG waveform itself can indicate the changes in BP. Detection and analysis of such features would not only lead towards a better understanding of the BCG methodology, but such information can also be leveraged for better and accurate assessment of BP using only unobtrusive BCG sensors.

In this paper, we hypothesize that an increase in BP, such as that caused by an isometric exercise, will induce high amplitude oscillations (ringing) after the J-peak in the BCG heartbeat. In order to assess these changes, data were collected from healthy subjects performing isometric hand-grip exercise which has been shown to modulate mean arterial pressure (MAP). An algorithm was designed to capture energy in the second half of BCG heartbeats during and after the exercise period.

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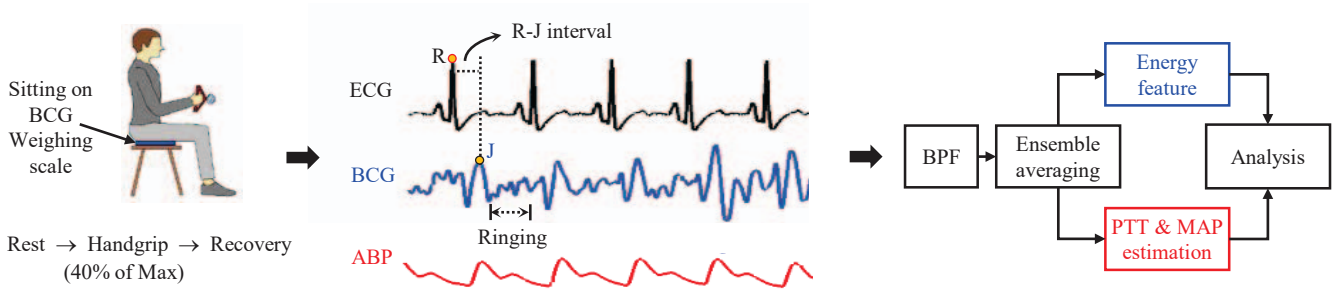

Fig. 1. Experimental setup. Data were collected from 14 healthy subjects. Each subject performed a hand-grip exercise while sitting on the BCG weighing scale in an upright posture. The latter half of the BCG signal (beyond the J-peak), which is composed of oscillations, was extracted for feature estimation and analysis. The ABP signals were used for calculating mean arterial pressure (MAP) and PTT.

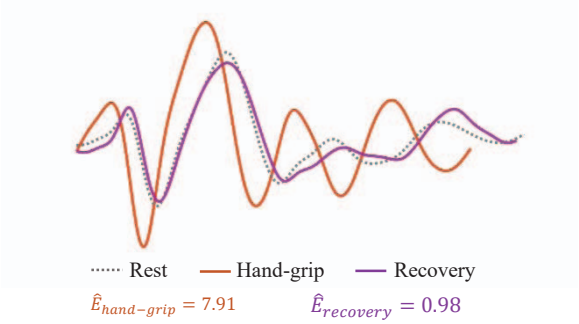

Fig. 2. Ensemble-averaged BCG heartbeats in the three phases of the protocol. The increase in oscillations beyond the J-peaks is clearly visible from the BCG heartbeat during the hand-grip phase of the protocol. The waves after the J-peak in the BCG heartbeat during the recovery phases look similar (in terms of peak-to-peak magnitude) to those in the resting phase BCG heartbeat.

The paper is organized as follows: the data collection protocol, hardware design and signal processing methods for analysis of the BCG signals are described in section II, followed by results, discussion and conclusion in sections III, IV and V, respectively.

II. Methods

A. Protocol

The data for the project were collected from 14 subjects under a protocol approved by the Institutional Review Board (IRB). Each subject was asked to sit for 60 seconds on a weighing scale modified to measure BCG signals using a custom analog amplifier. This was followed by 30 seconds of isometric contraction performed in the form of a hand-grip exercise to increase mean arterial pressure (MAP). Finally each subject was asked to recover for around 180 seconds while sitting on the scale as shown in Fig. 1 (left).

B. Hardware Design

The electrocardiogram (ECG) was measured using the BN-EL50 (BIOPAC Systems, Inc., Goleta, CA) and arterial blood pressure (ABP) was measured using the A2SYS Nexfin Monitor (Edwards Lifesciences, Irvine, CA) that used the volume clamping technique on one finger. The BCG data were collected with a modified-weighing scale [9]. All the signals were simultaneously recorded at $f_s = 2$ kHz and transmitted to the data acquisition system (MP150WSW, BIOPAC Systems, Inc., Goleta, CA).

C. Pre-Processing of Data

The ECG and BCG signals were filtered using finite-impulse response (FIR) band-pass filters (Kaiser window, pass-band $f_{pass} = 0.8 - 40$ Hz for the ECG and $f_{pass} = 0.8 - 20$ Hz for the BCG) while the ABP signals were low-pass filtered (FIR, Kaiser window, cut-off frequency $f_c = 20$ Hz) to preserve the dc value. The R-peaks in the ECG signal were detected with a threshold-based peak detector algorithm and manually verified to remove any spurious peaks. With R-peaks R_i as fiduciary points and l as the minimum R-R interval, $R_i + l$ portions / frames were extracted from the BCG and ABP signals of each subject. A collection of these frames, called an *ensemble*, were then averaged to obtain ensemble-averaged waveforms. All the frames extracted from the resting phase were averaged to obtain resting state ensemble-averaged waveforms for the BCG and ABP for each subject. The hand-grip and recovery portion of the recording for each subject were divided into 10-second intervals and all the frames in each 10-second interval were averaged to obtain ensemble-averaged traces. In the rest of the paper, we shall refer to these time-averaged waveforms as *heartbeats*.

D. Extraction of Features

The features extracted from the ABP and BCG heartbeats are explained in the following sub-sections.

1) Ringing Energy: In order to estimate the increase in the oscillations (ringing) portion of the BCG heartbeat beyond the K-peak (as shown in Fig. 2), the energy in that portion of the heartbeat was calculated using a simple polynomial fitting and envelope estimation method. First, the I-, J- and K-peaks were detected on the BCG heartbeat as shown in Fig. 3 (a). A portion w' was extracted from each BCG heartbeat ($w' =$ K-peak$-\alpha$ ms to end of the heartbeat, α = 12.5 ms in Fig. 3 (a)). The dc component present in w' was then eliminated by fitting and subtracting a second degree polynomial as shown in Fig. 3 (a). The dc component was removed to get rid of

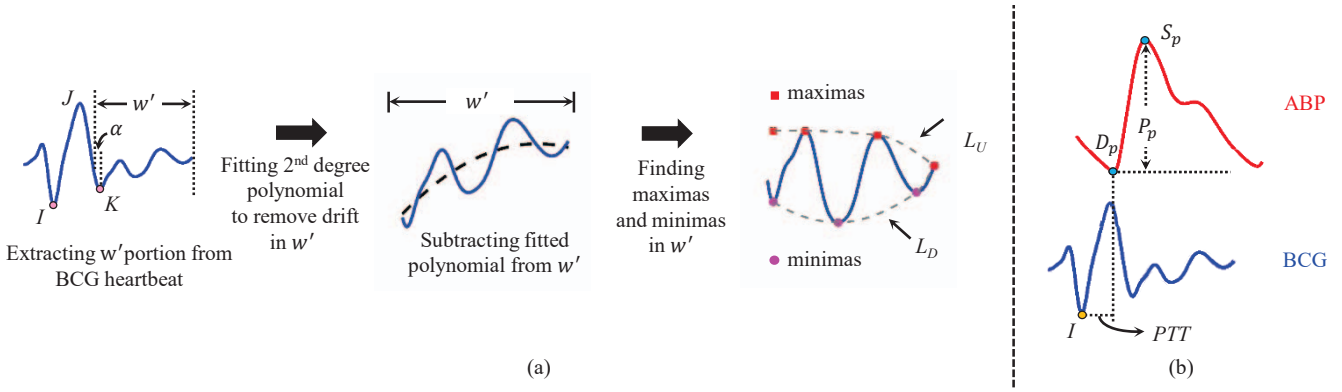

Fig. 3. (a) Estimation of ringing energy. w' ms portion after the J-peak was extracted from each BCG heartbeat. The dc component in the extracted signal was removed by fitting and subtracting a 2^{nd}-degree polynomial. The energy from the upper and lower envelopes of the extracted portion. (b) The S_p, D_p and P_p values were calculated from the ABP heartbeats. The PTT was estimated as the time difference between the foot (D_p) and I-peak of the BCG heartbeat.

any amplitude bias in calculation of energy. To estimate the energy in the resultant signal, the upper and lower envelopes (L_U and L_D) were estimated using cubic interpolation of the maxima and minima as shown in Fig. 3 (a). For equal lengths of L_U and L_D, the first maxima in L_U was arbitrarily chosen at K-peak position with the same amplitude value as the next detected maxima. The energy in w' was then estimated as

$$E = \frac{1}{N}\sum_{i=1}^{N}(L_U[i] - L_D[i])^2, \tag{1}$$

where N is the total number of samples. The ringing energy values from BCG heartbeats in the hand-grip and recovery phases for each subject were divided by the ringing energy from the corresponding resting heartbeat of that subject. The normalized values were denoted by $\widehat{E}$. The BCG heartbeats from one subject are shown in Fig. 2 with corresponding $\widehat{E}$ values.

2) PTT & MAP Calculation: Two features were extracted from the ABP heartbeats. In order to estimate MAP, the systolic (S_p) and diatolic (D_p) blood pressure values were detected as the peak and foot of the ABP heartbeat as shown in Fig. 3 (b). The difference between S_p and D_p is called pulse pressure (P_p). The MAP was then estimated as $MAP = D_p + P_p/3$. The MAP values estimated from the heartbeats during the hand-grip and recovery phases were normalized by corresponding resting phase values.

In order to estimate PTT, the I-peak in the BCG heartbeat was detected as the minima before the global J-peak. The time difference between the foot (D_p) of the ABP heartbeat and the I-peak of the BCG heartbeat was estimated as the PTT as shown in Fig. 3 (b). The inverse of the PTT values (PTT^{-1}) was used in the analysis. The PTT^{-1} values during the hand-grip and recovery phases were also normalized by resting phase values for each subject.

III. Results

The results for PTT^{-1} and MAP for one subject are shown in Fig. 4 (a). The corresponding BCG heartbeats during the hand-grip phase are shown in Fig. 4 (b). Since the hand-grip and recovery phase of the data was divided into 10-second intervals, 3 heartbeats are shown in Fig. 4 (b). The BCG heartbeats during recovery phase for the same subject are shown in Fig. 4 (c). It is evident from the figures that as MAP and PTT^{-1} increases during the isometric exercise, the peak-to-peak amplitude of the oscillations from K-peak onwards also increases.

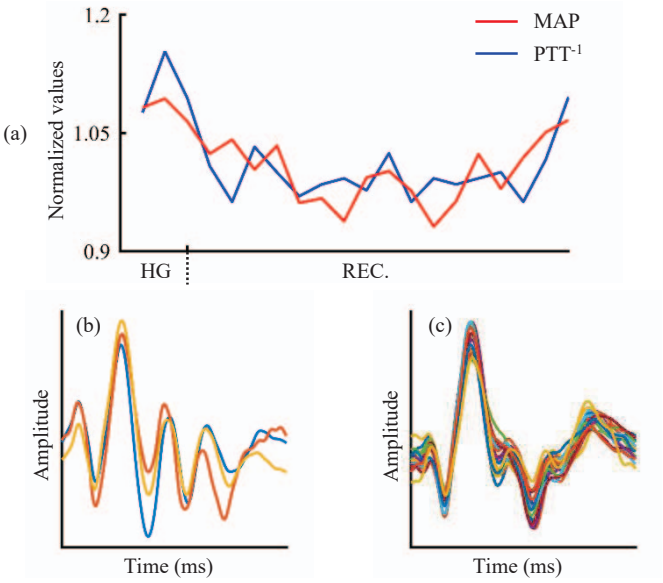

Fig. 4. (a) PTT^{-1} and MAP results for one subject during hand-grip (HG) and recovery (REC.) phases. (b) BCG heartbeats during the hand-grip phase. (c) BCG heartbeats during the recovery phase.

The values of normalized energy $\widehat{E}$ were estimated corresponding to the maximum and minimum values of normalized PTT^{-1} for each subject. Similarly, the $\widehat{E}$ were also obtained from maximum and minimum values for normalized MAP for each subject. The results are summarized in Fig. 5. There is a statistically significant difference between $\widehat{E}$ corresponding to maximum and minimum MAP for all subjects indicating that as the MAP increases due to isometric

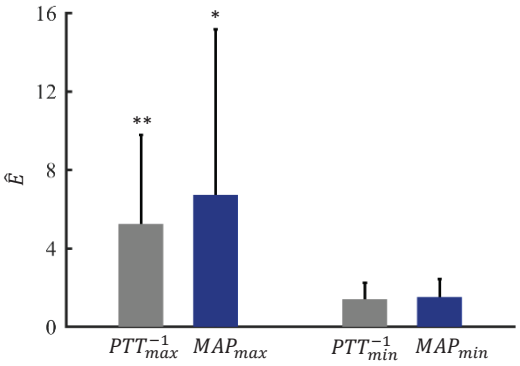

Fig. 5. Comparison of $\widehat{E}$ for maximum and minimum MAP and PTT^{-1} for all subjects. There is a statistically significant increase in $\widehat{E}$ for increase in MAP and PTT^{-1} ($* \rightarrow p < 0.05$, $** \rightarrow p < 0.01$)

exercie, the energy and amplitude of the oscillations (beyond the J-peak) in the BCG heartbeat also increase. A more significant result ($p < 0.01$) was obtained for $\widehat{E}$ comparison corresponding to maximum and minimum PTT^{-1}.

IV. Discussion

BCG is not as well understood as ECG, or other more commonly employed cardiovascular sensing modalities and, accordingly, the results of this paper are important for improving the understanding of this signal. The "after-waves" present in the BCG heartbeat, in particular—the waves following the J-peak—are poorly understood and have not been directly linked to any physiological phenomenon. In this paper, we were able to show that these "after-waves" are directly related with the MAP. Increase in MAP leads to an increase in the amplitude of these oscillations.

An important advantage of the BCG methodology is the fact that most of the BCG measuring sensors, e.g., modified-weighing scales, are already present in millions of homes across the world, making it ideal for continuous assessment of cardiovascular health at home. However, the number of cardiac markers which can be assessed using the BCG are still limited to systolic time intervals, more specifically only to PEP. Moreover, estimation of parameters related to BP, such as PTT, require to merge BCG with other sensing modalities. Hence, this paper lays an important foundation for leveraging the information present in the second half of the BCG heartbeat for assessment of BP. Future work should focus on using this information along with the proximal-timing reference provided by the BCG to directly obtain an accurate estimate of MAP from it.

One drawback of this study is the homogeneous nature of the data set used for analysis. Future studies should include data from subjects which are diagnosed with hypertension. At the same time, in order to further understand the morphology of BCG, future work should also include other physical and mental exercises which can modulate BP. Nevertheless, this study shows the potential of BCG methodology to monitor BP, and thus, can lead to an increase in the breadth of parameters measured by it.

V. Conclusion

In this paper, we present a novel method, based on polynomial fitting for energy estimation, to assess the changes in the BCG signals associated with the changes in BP. The results of this paper provide a powerful foundation to further assess the vibrations present in the BCG signal and hence use this information for monitoring markers related with cardiovascular health.

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